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Effect of triceps surae and quadriceps muscle fatigue on the mechanics of landing in stepping down in ongoing gait

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The aim of this study was to evaluate the effects of muscle fatigue of triceps surae and quadriceps muscles in stepping down in ongoing gait. We expected that the subjects would compensate for muscle fatigue to prevent potential loss of balance in stepping down. A total of 10 young participants walked over a walkway at a self-selected velocity to step down a height difference of 10-cm halfway. Five trials were performed before and after a muscle fatigue protocol. Participants performed two fatigue protocols: one for ankle muscle fatigue and another for knee muscle fatigue. Kinematics of and ground reaction forces on the leading leg were recorded. Fatigue did not cause a change in the frequency of heel or toe landing. Our results indicate that in stepping down fatigue effects are compensated by redistributing work to unfatigued muscle groups and by gait changes aimed at enhancing balance control, which was however only partially successful.

Practitioner Summary: Problems with transitions between levels in gait are common causes of falls. Fatigue can affect the ability to dissipate energy gained during the step down. The main finding was that young adults recruited unfatigued muscle groups and adjusted their gait pattern to compensate the fatigue effects during landing.

Keywords: muscle fatigue; stepping down; gait; balance control

Introduction

Stepping down an elevation in ongoing gait is a common task, such as when stepping off the sidewalk to cross a street. Problems with transitions between levels in gait, such as when stepping from a curb, are common causes of falls (Lord et al. 1993; Ivers et al. 1998; Tinetti, Doucette, and Claus 1995). Misplaced steps formed the third cause of falls and accounted for 12% of all cases (Berg, Alessio, and Mills 1997). To avoid a fall during landing a step, the leading leg must dissipate this kinetic energy at landing by means of eccentric contractions (van Dieën et al. 2008). Failure to do so, as occurs in unexpected stepping down, can cause a loss of balance (van Dieën et al. 2007).

Muscle fatigue, defined as a decrease in the force generating capacity of muscles (Edwards 1981), is a common phenomenon in the general population (Hancock and Desmond 2000) and is even more pronounced in aged and diseased populations (Vestergaard et al. 2009; Feasson et al. 2006). Muscle fatigue may impair motor control by increasing force variability and by limiting generation of fast corrective torques (Ribeiro, Mota, and Oliveira 2007; Lin et al. 2009). Fatigue has been reported to reduce stability of gait on level ground and to increase the risk of falls (Helbostad et al. 2007; Parijat and Lockhart 2008a, 2008b; Murdock and Hubley-Kozey 2012). At the same time, muscle fatigue appears to coincide with compensatory gait changes that enhance gait stability, such as an increased step width (Helbostad et al. 2007).

With respect to negotiating a curb, it was shown that fatigue increases the variability of placement of the trailing and leading limb relative to the curb, suggesting a loss of control and consequently an increased risk of making a misstep (Barbieri et al. 2013a). The effects of muscle fatigue on the step down are not known, but it can be surmised to affect the ability to dissipate energy gained during the step down. The strategy used to step down is dependent on characteristics of the environment such as the height of the step (van Dieën and Pijnappels 2009). In stepping down, the quadriceps muscle dissipates most of the energy at landing when a heel landing strategy is used, while the triceps surae muscle does so when a toe landing strategy is used (van Dieën et al. 2008). In the last stride before stepping down, quadriceps muscle fatigue coincided with a decreased muscle activity of both triceps surae and quadriceps muscles (Barbieri et al. 2013a). On the other hand, triceps surae muscle fatigue coincided with an increased muscle activity of the biceps femoris in the stance phase and rectus femoris in the swing phase (Barbieri et al. 2013a). These findings indicate differential effects of quadriceps and triceps surae muscle fatigue on the approach phase and possibly also on the landing phase in stepping down.

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Therefore, the aim of this study was to evaluate the effects of muscle fatigue of quadriceps and triceps surae muscles at landing in stepping down in ongoing gait. We induced quadriceps and triceps surae muscle fatigue, using a functional task, to guarantee ecological validity. While functional tasks do not cause isolated muscle fatigue in a single muscle group, tasks were selected to cause effects predominantly on either of these muscles. We expected that the subjects would compensate for muscle fatigue, to prevent a potential loss of balance at landing. We thus hypothesised an increase in the use of the heel landing strategy with ankle muscle fatigue and an increase in toe landing with knee muscle fatigue. Second, within the same strategy we hypothesised a shift of negative work from the fatigued muscles to non-fatigued muscles, for example with ankle muscle fatigue we expected a decrease in peak moments, peak negative power and negative work around the ankle and corresponding increases around the knee and hip. Third, we hypothesised a conservative strategy in stepping down with fatigue, evidenced by more negative work at landing (van Dieën et al. 2008) and a larger reduction of the forward angular and linear momenta (van Dieën et al. 2007). Finally, we hypothesised a more conservative gait pattern with fatigue, reflected in an increased step width and reduced step duration (Hak et al. 2012), and a larger step length in the step down (van Dieën et al. 2007).

Material and methods

Participants

After signing informed consent, 10 health volunteers (5 men and 5 women – age: 27.60 ± 2.79 years; weight: 69.21 ± 10.76 kg; height: 1.81 ± 0.08 m) participated in the study that had been approved by the local ethics committee. Participants reported no history of musculoskeletal disorders, which could affect performance of the experimental tasks, within the past 12 months.

Experimental design

Data collection was performed on two days with a one-week interval (Figure 1). On the first day, the participants were familiarised with the study protocol and performed stepping down tasks before and after the first of two fatigue protocols. The two fatigue protocols aimed to induce fatigue of: (1) the quadriceps muscle and (2) the triceps surae muscle. The order of the fatigue protocols was balanced over subjects, to avoid order effects. Participants were instructed not to perform intense physical activity within 48 hours before data collection.

Fatigue protocols

To induce quadriceps muscle fatigue, the participants performed a repeated sit-to-stand task from a standard chair (40- cm high, 40- cm wide, 35- cm deep) without arm supports, with arms across the chest, at frequency of 0.5 Hz controlled by a metronome (Helbostad et al. 2007). The instruction given to the participants was: stand up to an upright position with your knees fully extended, then sit back down and repeat this at the beat of the metronome until you can no longer perform the task.

To induce triceps surae muscle fatigue, the participants performed a repeated standing calf raise exercise (Berger, Regueme, and Forestier 2010) at 0.5 Hz. During the task, participants were allowed to place their hands against the wall to avoid balance loss. The instruction given to the participants was: lift your heels off the ground as high as possible then return to the starting position and repeat this at the beat of the metronome until you can no longer perform the task. Preceding both fatigue protocols, participants performed some attempts to adapt to the task and the frequency.

The fatigue protocol was stopped when the participant indicated that he or she was unable to continue, or when the movement frequency fell below and remained below 0.5 Hz after encouragement, or after 30 min. No rest period was

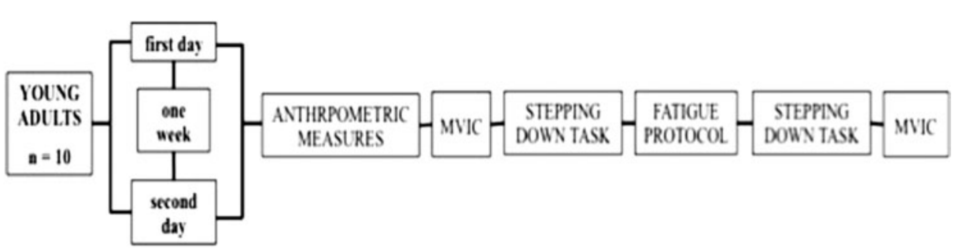

Figure 1. Experimental design.
Note: MVIC = maximal voluntary isometric contractions.

allowed between trials and testing was started as soon as possible after the fatigue protocol and the time between the fatigue protocol and the gait trials (<3 min) was expected not to allow for full recovery (Parijat and Lockhart 2008b). Isometric force measurements were used to confirm that the muscle was fatigued, measuring the loss of strength (Barbieri et al. 2013a; Katsiaras et al. 2005; Petrella et al. 2005; Stackhouse et al. 2001). Finally, the endurance time during the fatigue protocol was recorded.

Isometric force measurements

Maximal voluntary forces of the quadriceps and triceps surae muscles of the right leg were measured using custom-made dynamometers (Pijnappels et al. 2008). Subjects performed two attempts of maximal voluntary isometric contractions (MVIC) before and after the fatigue protocol, with at least 2 min rest between attempts. Subjects were asked to generate maximal isometric knee or ankle extension for 3–4 s. For knee extension, the participants were seated in a backward inclined chair, with the hip joint at 90° (180° is full extension) and knee joint at 120° (180° is full extension) and the lower leg tightly strapped to a strain gauge transducer (KAP, E/200 Hz, Bienfait BV Haarlem, The Netherlands) placed ~25 cm distal from the knee joint, which measured the force exerted at the shin. For ankle extension, the participants were seated in an upright chair with the knee joint flexed at 90° and the ankle joint set at 20° dorsiflexion, and the forefoot positioned on a 10 cm $\times$ 10 cm force transducer (AMTI M3-1000, Watertown, USA) that was mounted in the push-off platform. Forces were sampled at 1 kHz. The means of the two attempts before and after muscle fatigue were calculated for each participant.

Stepping task

Before and after the fatigue protocol, the participants performed five trials of stepping down a 10-cm elevation. Subjects walked at a self-selected speed over a raised platform of 10 m and halfway they stepped down onto the lower part of the platform. At the 10-cm height difference, individuals are expected to alternate between toe and heel landing strategies (van Dieën and Pijnappels 2009). Before the actual experiment, the participant was accustomed to the testing environment. No rest period was allowed between trials and testing was started as soon as possible after the fatigue protocol. The starting point was adjusted individually to make sure that subjects stepped down with the right leg.

The strategy used to step down was determined from visual inspection of the videotapes recorded during the experiment by the same observer for all trials. Strategies were classified as toe landing or heel landing based on whether the heel or forefoot/toes first touched the ground.

An optical system with 3×3 cameras (Optotrak Northern Digital Inc., Waterloo, ON, Canada) was used to record kinematics at a rate of 50 samples/s. Neoprene bands, with clusters of three infrared Light Emitting Diodes, were placed on the pelvis (level of the posterior superior iliac spines), bilateral thigh, shank and heels. Cluster positions were related to anatomical landmarks based on separate pointer measurements (Cappozzo et al. 1995). Ground reaction forces were recorded at 1000 samples/s using two 1.0×1.0 m force plates, placed in the walking platform just before and after the 10-cm height difference. The kinematic data and ground reaction forces were synchronised and offline reduced to 50 samples/s using a running average. Kinematic and force plate data were filtered with a bidirectional second order low-pass Butterworth filter with a cut-off frequency of 20 Hz. The analysis focused on the first stance phase at the lower level (leading leg). Sagittal plane knee, ankle and hip joint angles, angular velocities, moments, power and work of the leading (right) leg were estimated using a 3D linked-segment model (Kingma et al. 1996) and peak values occurring within 200 ms after ground contact of the leading leg at the lower level were determined (van Dieën et al. 2008). Changes in linear and angular momentum after landing were calculated from the areas under the curve of ground reaction force and the external moment about the body centre of mass, respectively (van Dieën et al. 2007). In addition, the step length, step width, step duration, step velocity of the landing step and the heel clearance during stepping down were analysed. The heel clearance was calculated as the distance of the calcaneus landmark to the step.

Statistics

The dependent variables of interest were statistically analysed with SPSS 18.0 for Windows® ($\alpha < 0.05$). The data were normally distributed, as verified by the Shapiro–Wilk test. First, to verify the similarity of the data in the unfatigued conditions on both days, the variables of interest were evaluated by paired sample *t*-tests. The same statistical test was used to compare the endurance times between fatigue protocols, and the MVIC values before and after fatigue. The frequencies at which a heel or toe landing strategies were used by each subject were compared between the conditions before and after fatigue, using Wilcoxon tests. The dependent variables during stepping down were compared using repeated measures multivariate ANCOVAs, with fatigue (before and after fatigue) as main factor and landing strategy (heel or toe landing) as covariate. Separate multivariate ANCOVAs were performed for each fatigue protocol and for joint kinetics, for the overall mechanics of the step down and for spatial and temporal gait parameters. When the ANCOVA pointed out significant differences, Tukey univariate tests were carried out, as proposed by Zar (1999).

Results

The participants performed the ankle fatigue protocol (Table 1) longer than the knee fatigue protocol ($t_9 = 4.27$, $p < 0.002$). Four participants performed for 30 min in the ankle fatigue protocol, while none performed for 30 min in the knee fatigue protocol. After the fatigue protocol, the MVIC had decreased both in the ankle (15.5%; $t_9 = 5.20$, $p < 0.001$) and in the knee (3.2%; $t_9 = 3.46$, $p < 0.007$) fatigue protocol.

Overall, the participants preferred stepping down the step using a heel landing. In contrast with our hypothesis, neither protocol caused significant differences in frequencies of the landing strategies between before and after muscle fatigue ($p > 0.28$; Figure 2).

There were no significant differences in the variables of interest between trials before the ankle and knee muscle fatigue protocols. The results are illustrated by a comparison of a mean values and standard deviations before and after muscle fatigue (Figures 3–4). Marked differences between before and after muscle fatigue for both fatigue protocols occurred in the joint moment and power. Group averages were presented in the tables.

For the ankle fatigue protocol (Table 2), the ANCOVA on joint kinetics revealed main effects of fatigue (Wilks' $\lambda = 0.50$, $F_{9,40} = 4.34$, $p < 0.001$) and interactions between fatigue and landing strategy (Wilks' $\lambda = 0.61$, $F_{9,40} = 2.83$, $p < 0.01$). In line with our hypothesis, a redistribution of work towards unfatigued muscle groups was indicated by main effects of fatigue on peak hip moments and negative work around the hip (Figure 3). In toe landing, a shift away from the ankle muscles was indicated by significant effects on peak ankle and knee moments, peak ankle power and negative work around the knee, as well as a trend in negative work around the ankle. For heel landing, only a significant decrease in negative work around the knee and an increase in peak power around the hip were observed.

The ANCOVA on the overall mechanics of landing revealed a main effect of ankle muscle fatigue (Wilks' $\lambda = 0.73$, $F_{4,45} = 4.02$, $p < 0.007$) and an interaction effect of fatigue and strategy (Wilks' $\lambda = 0.80$, $F_{4,45} = 2.71$, $p < 0.04$). The interaction was caused by the total negative work, with a significant increase with fatigue in toe landing only (Table 2). The reduction in angular momentum at landing, however, decreased with fatigue irrespective of the landing strategy, while the changes in linear momentum were not affected by fatigue (Table 2).

The ANCOVA on spatial and temporal gait parameters revealed no main effect of fatigue (Wilks' $\lambda = 0.73$, $F_{6,43} = 2.52$, $p < 0.03$), but an interaction effect of fatigue and strategy (Wilks' $\lambda = 0.71$, $F_{6,43} = 2.79$, $p < 0.02$) was found. This effect was accounted for by an interaction effect on step width, with a significant increase in step width occurring in toe landing only (Table 2).

Knee muscle fatigue had main effects on joint kinetics, overall mechanics of landing and on the spatial and temporal parameters only (Wilks' $\lambda = 0.43$, $F_{9,40} = 5.89$, $p < 0.001$; Wilks' $\lambda = 0.80$, $F_{4,45} = 2.81$, $p < 0.03$; Wilks' $\lambda = 0.59$, $F_{6,43} = 4.84$, $p < 0.01$, respectively) and no interactions with strategy (Wilks' $\lambda = 0.37$, $F_{9,40} = 7.37$, $p = 0.08$; Wilks' $\lambda = 0.85$, $F_{9,40} = 1.89$, $p < 0.12$; Wilks' $\lambda = 0.67$, $F_{6,43} = 3.50$, $p < 0.07$, respectively).

In line with our hypothesis, the energy absorption was increased around the ankle when the knee muscles were fatigued. In addition, peak power around the hip increased after muscle fatigue (Figure 4). Furthermore, increases in total negative work and step width and a decrease in step duration (Table 3) were in line with the use of a more conservative strategy in the fatigued condition.

Discussion

We hypothesised that ankle and knee muscle fatigue would cause changes in kinematics and kinetics of stepping down that would reflect a shift of work away from the fatigued muscle groups. In contrast with this hypothesis, we found no change in

Table 1. Mean values and standard deviations of the endurance times and maximal voluntary isometric contractions (MVIC) before and after fatigue.

	Ankle fatigue	Knee fatigue
Endurance time(s)	1179 ± 584.32[a]	405.80 ± 435.15
MVIC		
Before fatigue	127.49 ± 31.32[b]	250.37 ± 11.49[b]
After fatigue	107.49 ± 28.97	242.39 ± 9.34

[a] Significant differences between ankle and knee muscle fatigue.
[b] Significant differences between the unfatigued and fatigued conditions.

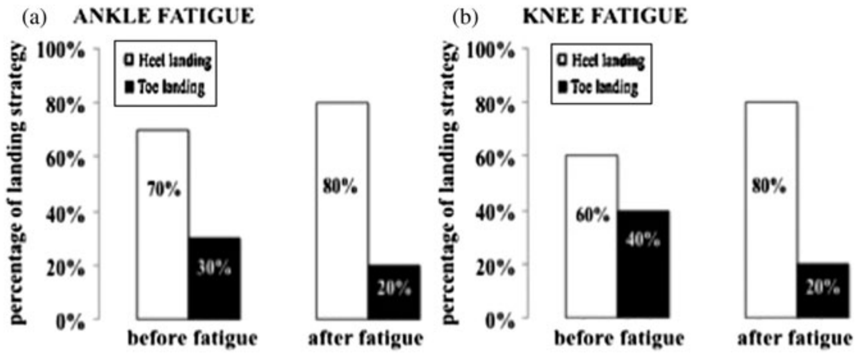

Figure 2. Use of heel and toe landing as a percentage of all trials before and after fatigue, for (a) ankle muscle fatigue and (b) knee muscle fatigue.

the frequency of heel or toe landing after the fatigue protocols. However, changes in joint kinetics were in line with the expected shift in work away from the fatigued muscles (either ankle or knee muscles) towards (relatively) unfatigued muscles (hip and either knee or ankle muscles). Furthermore, we hypothesised that changes in the mechanics of stepping down with fatigue would reflect a strategy to reduce the risk of balance loss. Stepping down coincides with increases in forward linear and angular momenta, which are dealt with at landing (van Dieën et al. 2007, 2008). Increases in the negative work in the leading leg at landing, found in most cases, are in line with this hypothesis. However, the reductions in the linear momenta at landing were not changed and in case of ankle muscle fatigue the reduction of angular momentum at landing was even decreased. This suggests that subjects were not successful at increasing control over the body's momentum at stepping down. Overall, subjects increased step width, which may reflect an attempt to enhance balance control, but they did not adjust step length (in both muscle fatigue) and step duration (only for ankle muscle fatigue).

The fatigue protocols used in the present study induced a significant reduction of muscle strength, especially the ankle muscle fatigue protocol (reduction of 15.7%). The longer endurance time in the ankle fatigue protocol may account for the larger reduction in strength. The reduction in strength could reduce joint moments and hence absorption of the kinetic energy gained (Prilutsky et al. 1996; Devita, Blankenship-Hunter, and Skelly 1992) and joint stabilisation at landing

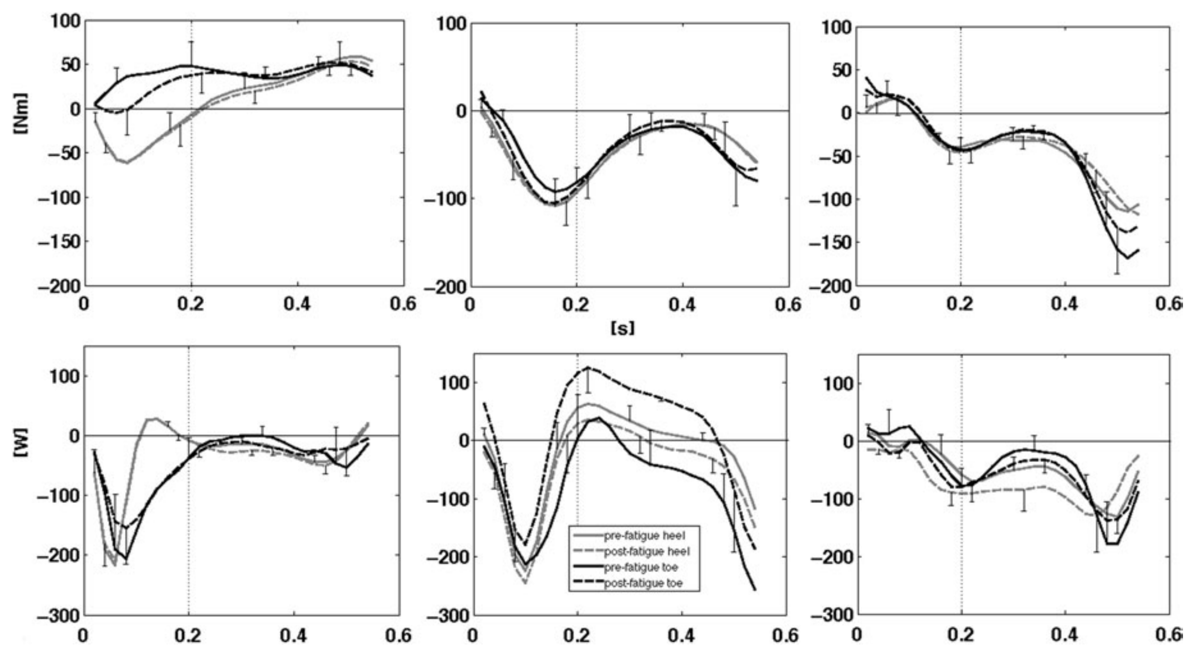

Figure 3. Mean values and standard deviations of joint moments (top panels) and power (lower panels) before (black line) and after (gray line) ankle muscle fatigue for heel (solid line) and toe (line with circle) landing strategy.
Note: Left panels are ankle moment and power, middle panels are knee moment and power and right panels are hip moment and power. Vertical dashed line indicates the first 200 ms after ground contact of the leading leg at the lower level.

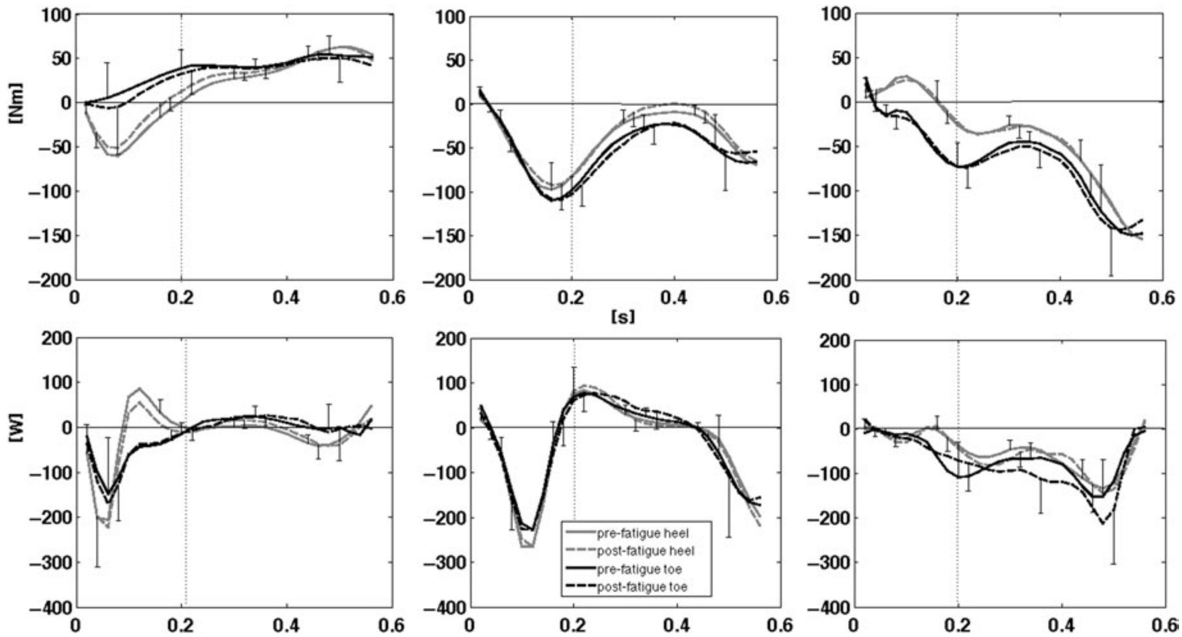

Figure 4. Mean values and standard deviations of joint moments (top panels) and power (lower panels) before (black line) and after (gray line) knee muscle fatigue for heel (solid line) and toe (line with circle) landing strategy.
Note: Left panels are ankle moment and power, middle panels are knee moment and power and right panels are hip moment and power. Vertical dashed line indicates the first 200 ms after ground contact of the leading leg at the lower level.

(Murdock and Hubley-Kozey 2012; Granacher et al. 2010; Vila-Chã et al. 2011; Lewek, Rudolph, and Snyder-Mackler 2004) through eccentric actions. Our results indicated that after ankle fatigue eccentric contractions of the knee muscles contributed more to reducing the kinetic energy that is gained in stepping down and vice versa after knee muscle fatigue, albeit without the kinematic strategy changing completely from toe to heel landing or vice versa. With ankle muscle fatigue, also the hip muscle contribution to negative work was increased and a similar trend was seen with knee muscle fatigue.

After the muscle fatigue protocols, the participants widened the base of support, which has also been found in individuals walking on level ground and crossing obstacles (Barbieri et al. 2013b; Parijat and Lockhart 2008a, 2008b; Helbostad et al. 2007). Increased step width provides a larger margin of safety in medio-lateral control of the body's centre of mass (Hak et al. 2012; Hof et al. 2007). In addition, with knee muscle fatigue the participants reduced step duration, which also facilitates balance control in the medio-lateral direction (Hof et al. 2007; Hof, Gazendam, and Sinke 2005). An increase in step length would facilitate control of the angular momentum gained in stepping down (van Dieën et al. 2007) and hence control of balance in the forward direction. Such a change was not observed, possibly because it would conflict with a reduction in step duration.

Participants did increase the total energy absorption at landing, when the knee muscles were fatigued and when the ankle muscles were fatigued, although in the latter case only when a toe landing strategy was used, in line with a more conservative strategy. This suggests that participants tended to overcompensate for the effects of fatigue to enhance control over stepping down. However, effects were small and the reduction in linear momentum at landing was not increased with fatigue. Moreover, angular momentum was even reduced less when the ankle muscles were fatigued. The participants did not change step length and heel clearance after muscle fatigue in both protocols. Heel clearance is the most obvious indicator of the risk of tripping in an uneven gait task (Patla and Rietdyk 1993). High heel clearance would be a safe strategy (Patla and Rietdyk 1993; Lu, Chen, and Chen 2006; Silva, Barbieri, and Gobbi 2011). Moreover, increasing step length at stepping down facilitates the reduction of the angular momentum gained (van Dieën et al. 2008). While step length and heel clearance were not increased, we have previously shown that variability of heel clearance increased with fatigue (Barbieri et al. 2013a). These findings combined may suggest the limited control of heel clearance and related to that the control of angular momentum. The effect on angular momentum, however, appears minor from the point of view of fall risk, since a comparison of expected and unexpected stepping down showed much larger effects on the reduction of the angular momentum at landing, without falls occurring (van Dieën et al. 2008). Nevertheless, this implies that subjects were only partially successful at enhancing balance control in the presence of fatigue.

Table 2. Mean values and standard deviations of parameters characterising the kinetics during landing (within 200 ms of landing) and spatial-temporal parameters during the step of stepping down before and after ankle muscle fatigue according to landing strategy (toe and heel landing).

	Toe landing		Heel landing			
	Before fatigue	After fatigue	Before fatigue	After fatigue	Effect of fatigue	Interaction of factors
Joint kinetics						
Peak ankle moment (Nm)	47.05 ± 16.05[a]	37.37 ± 12.92	−61.78 ± 13.14	−60.06 ± 12.75	0.07	0.009
Peak knee moment (Nm)	−94.46 ± 25.47[a]	−105.81 ± 31.84	−108.86 ± 30.27	−109.37 ± 32.36	0.06	0.05
Peak hip moment (Nm)	−42.76 ± 21.01	−44.08 ± 16.54	−39.97 ± 11.45	−48.79 ± 23.12	0.04	0.12
Peak negative ankle power (W)	−205.99 ± 90.12[a]	−153.24 ± 38.81	−216.64 ± 72.43	−208.59 ± 43.05	0.01	0.05
Peak negative knee power (W)	−212.94 ± 82.91	−180.98 ± 51.71	−224.76 ± 71.34	−245.79 ± 104.65	0.71	0.08
Peak negative hip power (W)	−77.87 ± 51.12	−80.12 ± 31.13	−59.50 ± 26.69[a]	−91.13 ± 30.83	0.03	0.009
Negative work ankle (J)	−18.46 ± 8.56	−12.57 ± 4.32	−18.46 ± 8.56	−15.14 ± 3.39	0.07	0.08
Negative work knee (J)	−17.05 ± 5.96[a]	−25.86 ± 15.83	−22.51 ± 10.02[a]	−18.93 ± 8.24	0.07	0.001
Negative work hip (J)	−6.30 ± 4.477	−7.53 ± 2.82	−8.93 ± 6.56	−10.50 ± 6.06	0.05	0.72
Overall landing mechanics						
Negative work total (J)	−41.81 ± 16.53[a]	−48.53 ± 18.65	−43.64 ± 10.75	−42.02 ± 10.29	0.06	0.002
Negative angular momentum (Nms)	−8.46 ± 3.61	−7.03 ± 3.21	−6.63 ± 5.37	−5.27 ± 3.26	0.02	0.95
Negative horizontal momentum (Ns)	−26.94 ± 7.27	−25.48 ± 5.46	−26.78 ± 6.88	−25.79 ± 5.62	0.45	0.06
Vertical momentum (Ns)	217.23 ± 30.73	213.95 ± 43.62	233.49 ± 62.23	229.17 ± 52.86	0.95	0.92
Spatial-temporal parameters						
Step length (cm)	76.74 ± 8.21	77.28 ± 9.71	80.56 ± 11.22	79.28 ± 9.71	0.69	0.23
Step width (cm)	10.72 ± 4.22[a]	14.92 ± 3.01	13.77 ± 3.30	13.79 ± 3.92	0.009	0.009
Step duration (s)	0.53 ± 0.04	0.52 ± 0.04	0.58 ± 0.08	0.56 ± 0.06	0.28	0.54
Step velocity (cm/s)	146.72 ± 21.37	148.95 ± 10.55	136.18 ± 22.28	141.32 ± 18.31	0.10	0.51
Heel clearance (cm)	13.89 ± 1.38	13.33 ± 3.12	11.15 ± 2.09	10.85 ± 1.97	0.32	0.76

[a] Significant differences between the unfatigued and fatigued conditions.

Table 3. Mean values and standard deviations of parameters characterising the kinetics during landing (within 200 ms of landing) and spatial-temporal parameters during the step of stepping down before and after knee muscle fatigue.

	Knee fatigue		
	Before fatigue	After fatigue	Effects of fatigue
Joint kinetics			
Peak ankle moment (Nm)	−20.44 ± 53.97	−23.23 ± 55.35	0.32
Peak knee moment (Nm)	−100.12 ± 26.13	−98.82 ± 23.25	0.81
Peak hip moment (Nm)	−43.73 ± 26.20	−41.72 ± 34.53	0.83
Peak negative ankle power (W)	−198.13 ± 64.28	−203.61 ± 76.21	0.54
Peak negative knee power (W)	−269.39 ± 97.67	−264.61 ± 96.05	0.90
Peak negative hip power (W)	−63.61 ± 26.33[a]	−51.15 ± 24.04	0.001
Negative work ankle (J)	−13.63 ± 4.49[a]	−15.01 ± 6.10	0.05
Negative work knee (J)	−19.67 ± 10.41	−21.06 ± 914	0.13
Negative work hip (J)	−5.23 ± 3.83	−5.94 ± 5.16	0.09
Overall landing mechanics			
Negative work total (J)	−38.54 ± 15.29[a]	−42.02 ± 14.06	0.01
Negative angular momentum (Nms)	−6.22 ± 2.71	−6.40 ± 2.67	0.87
Negative horizontal momentum (Ns)	−26.32 ± 5.71	−26.33 ± 5.78	0.90
Vertical momentum (Ns)	220.35 ± 45.43	226.65 ± 48.87	0.06
Spatial-temporal parameters			
Step length (cm)	77.48 ± 8.70	77.39 ± 7.95	0.82
Step width (cm)	12.71 ± 5.05[a]	14.02 ± 5.41	0.01
Step duration (s)	0.54 ± 0.05[a]	0.52 ± 0.04	0.005
Step velocity (cm/s)	143.44 ± 20.01	144.96 ± 14.48	0.29
Heel clearance (cm)	11.78 ± 2.10	11.87 ± 2.22	0.83

[a] Significant differences between the unfatigued and fatigued conditions.

Some limitations of this study are evident. First, we did not control the level of physical activity of participants, which could be a factor mediating the effects of muscle fatigue. However, a previous study on obstacle crossing did not show such effects (Barbieri et al. 2013b). Since participants performed 10 trials of stepping down in the first day and 10 more trials after one week, the results could be affected by practice. However, we controlled for this by balancing the order of the fatigue protocols over subjects. Although the fatigue protocols were designed to induce fatigue predominantly in the quadriceps and triceps surae muscles respectively, it is important to consider that other muscles are involved in and may be fatigued by the functional task. Finally, only young healthy adults were tested, which limits generalisation to aged and diseased populations.

In conclusion, our hypotheses were in part accepted. In line with our hypotheses, young adults used strategies to enhance the balance control in stepping down an expected elevation, such as a shift in work away from the fatigued muscles to unfatigued muscles, increases in the negative work in the leading leg at landing and an increase in step width. However, in contrast with our hypotheses, young adults did not change the frequency of heel or toe landing strategies after both fatigue protocols, and were not successful at increasing control over the body's momentum at stepping down. Therefore, we concluded that in stepping down with local leg muscle fatigue, young healthy adults compensated fatigue effects by recruiting unfatigued muscle groups and adjusting their gait pattern to enhance balance control.

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